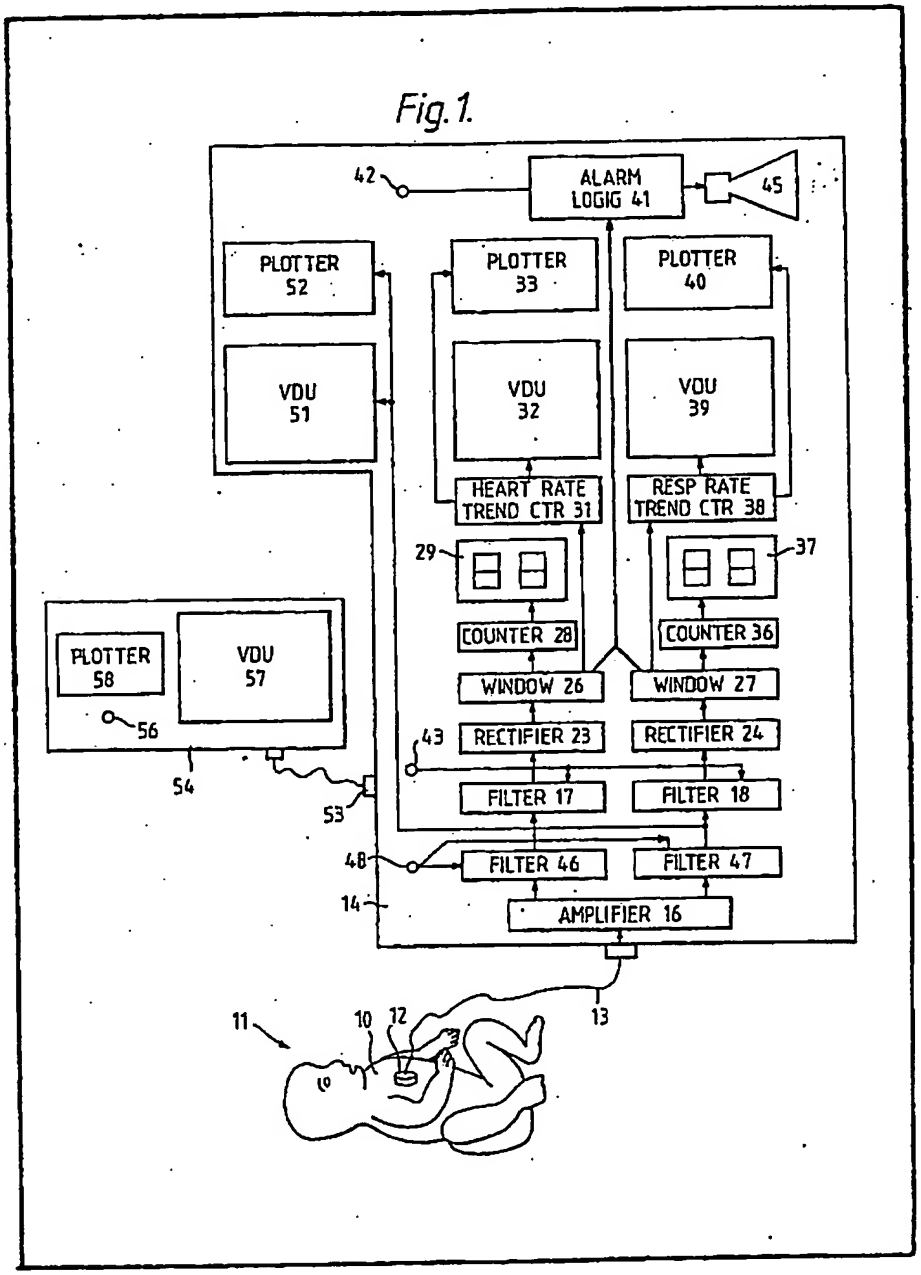
(12) UK Patent Application (19) GB (11) 2 129 991 A

(21) Application No 8329057
(22) Date of filing 31 Oct 1983
(30) Priority data
(31) 8230983
(32) 29 Oct 1982
(33) United Kingdom (GB)
(43) Application published
23 May 1984
(51) INT CL³
G08B 3/00
(52) Domestic classification
G4N EE
(56) Documents cited
None
(58) Field of search
G4N
(71) Applicant
The London Hospital Medical College, (United Kingdom), Turner Street, Whitechapel, London, E1 2AD
(72) Inventors
Neil McLellan, T. G. Barnett
(74) Agent and/or address for service
Brookes & Martin, 52/54 High Holborn, London, WC1V 6SE

(54) Cardio respiratory monitor

(57) A sound detector is placed on the body, the sound detected being divided into a first frequency component which relates to cardiac function and a second frequency component which relates to respiratory function. These signals are analysed and monitored.

Fig. 1.



GB 2 129 991 A

Fig. 1.

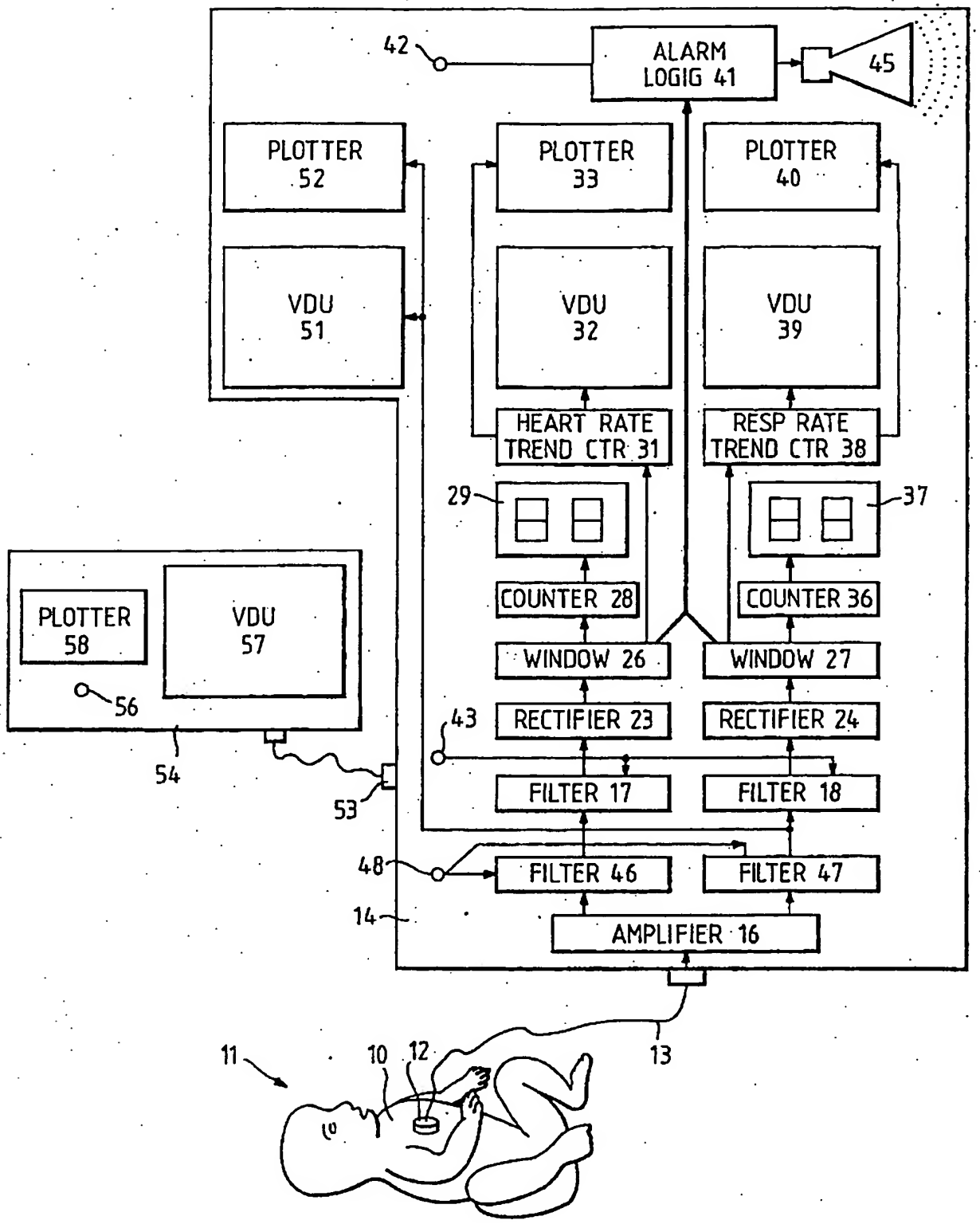

42
ALARM LOGIG 41
45
PLOTTER 52
PLOTTER 33
PLOTTER 40
VDU 51
VDU 32
VDU 39
HEART RATE TREND CTR 31
RESP RATE TREND CTR 38
29
37
COUNTER 28
COUNTER 36
WINDOW 26
WINDOW 27
RECTIFIER 23
RECTIFIER 24
43
FILTER 17
FILTER 18
48
FILTER 46
FILTER 47
14
AMPLIFIER 16
PLOTTER 58
VDU 57
56
54
53
13
11
10
12

Fig.2.

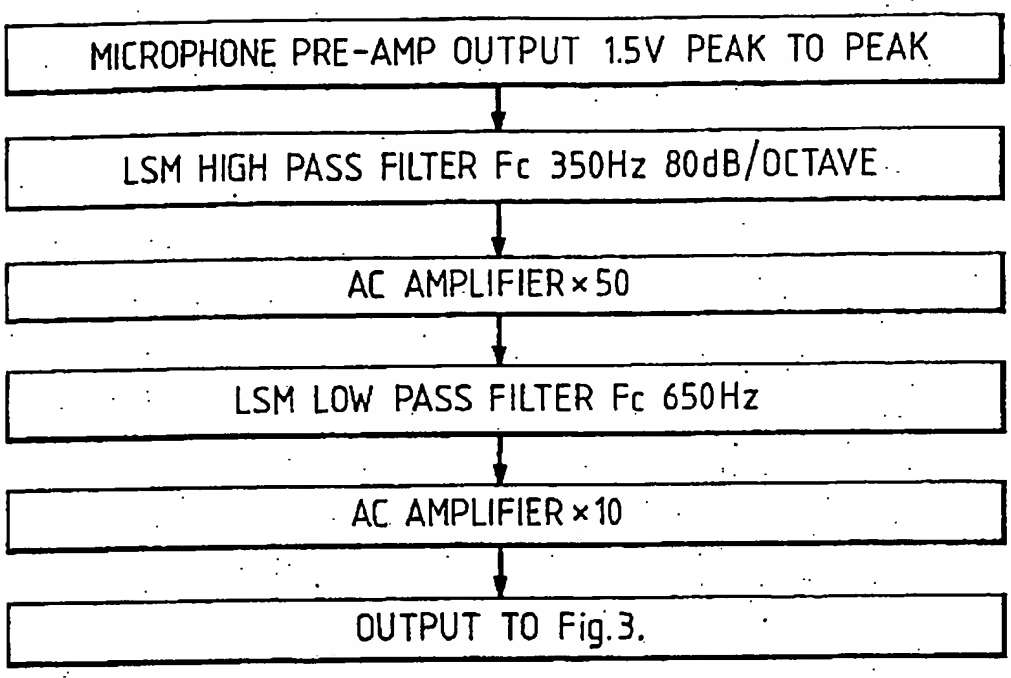

MICROPHONE PRE-AMP OUTPUT 1.5V PEAK TO PEAK
LSM HIGH PASS FILTER Fc 350Hz 80dB/OCTAVE
AC AMPLIFIER×50
LSM LOW PASS FILTER Fc 650Hz
AC AMPLIFIER×10
OUTPUT TO Fig.3.

Fig.3.

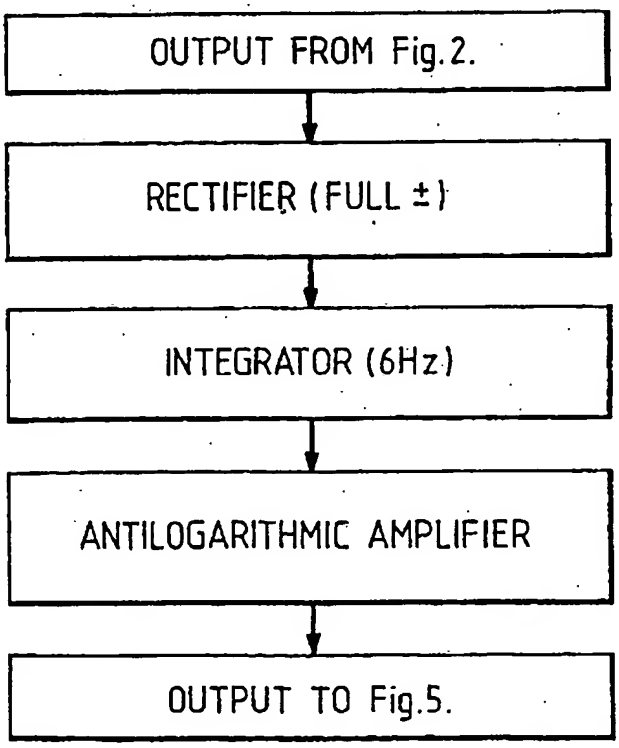

OUTPUT FROM Fig.2.
RECTIFIER (FULL ±)
INTEGRATOR (6Hz)
ANTILOGARITHMIC AMPLIFIER
OUTPUT TO Fig.5.

Fig.4A.

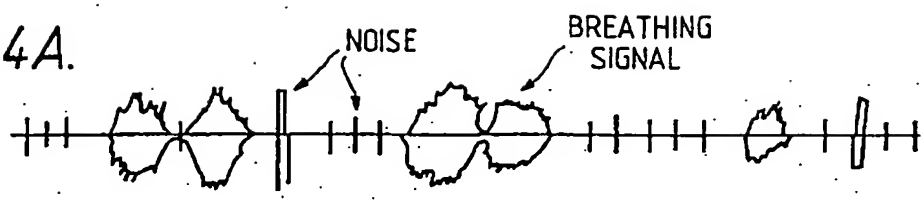


Fig.4B

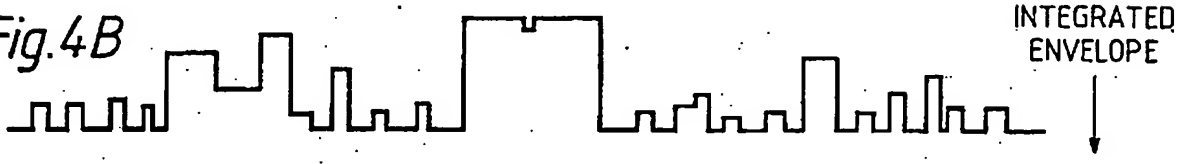
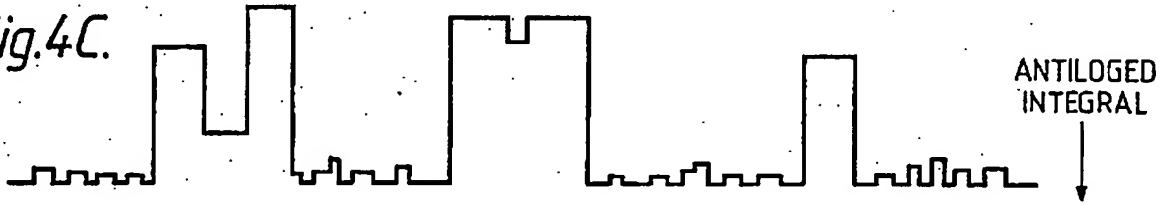


Fig.4C.



NOW SET THE VOLTAGE LIMIT 10% ABOVE THE NOISE FLOOR WITH THE VOLTAGE-LIMIT MODULE:-

Fig.4D.

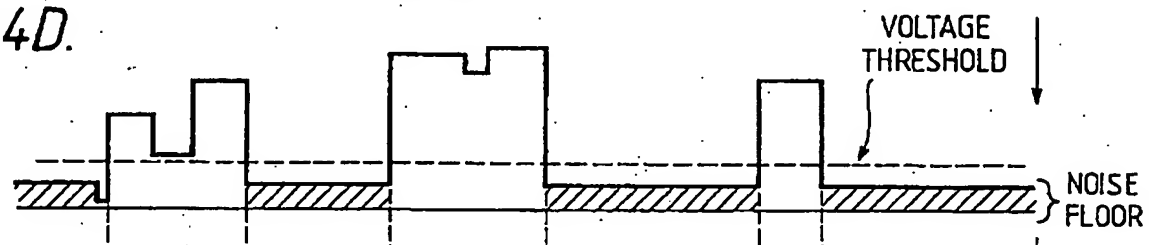
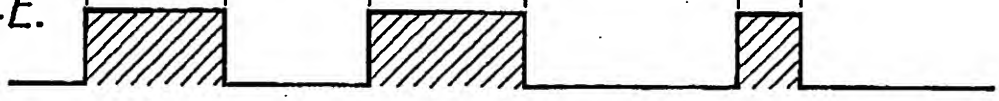


Fig.4E.



PULSE OUTPUT FROM VOLTAGE LIMIT MODULE

Fig.5.

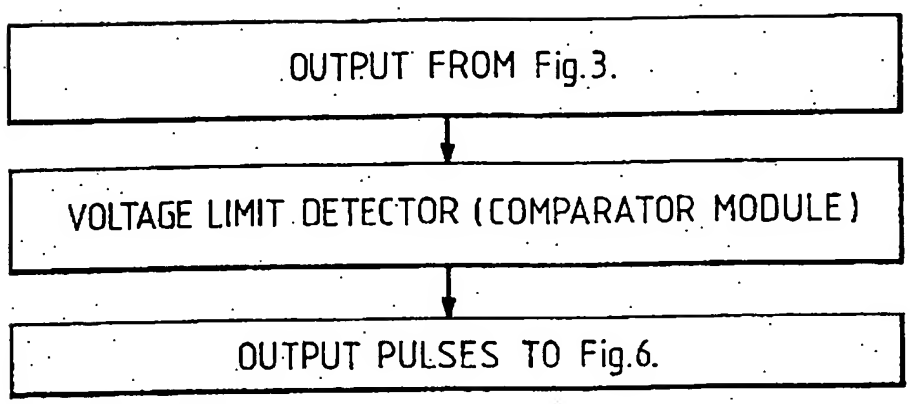
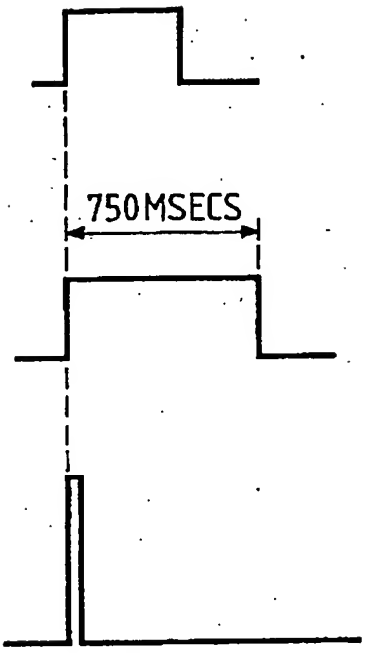

OUTPUT FROM Fig.3.
VOLTAGE LIMIT DETECTOR (COMPARATOR MODULE)
OUTPUT PULSES TO Fig.6.

Fig.6.


750MSECS

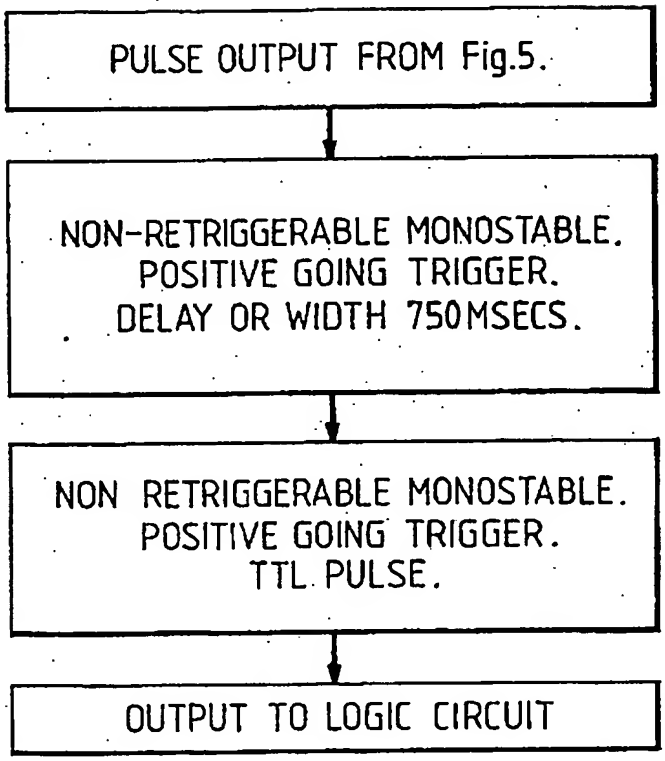

PULSE OUTPUT FROM Fig.5.
NON-RETRIGGERABLE MONOSTABLE.
POSITIVE GOING TRIGGER.
DELAY OR WIDTH 750MSECS.
NON RETRIGGERABLE MONOSTABLE.
POSITIVE GOING TRIGGER.
TTL PULSE.
OUTPUT TO LOGIC CIRCUIT

Fig.7.

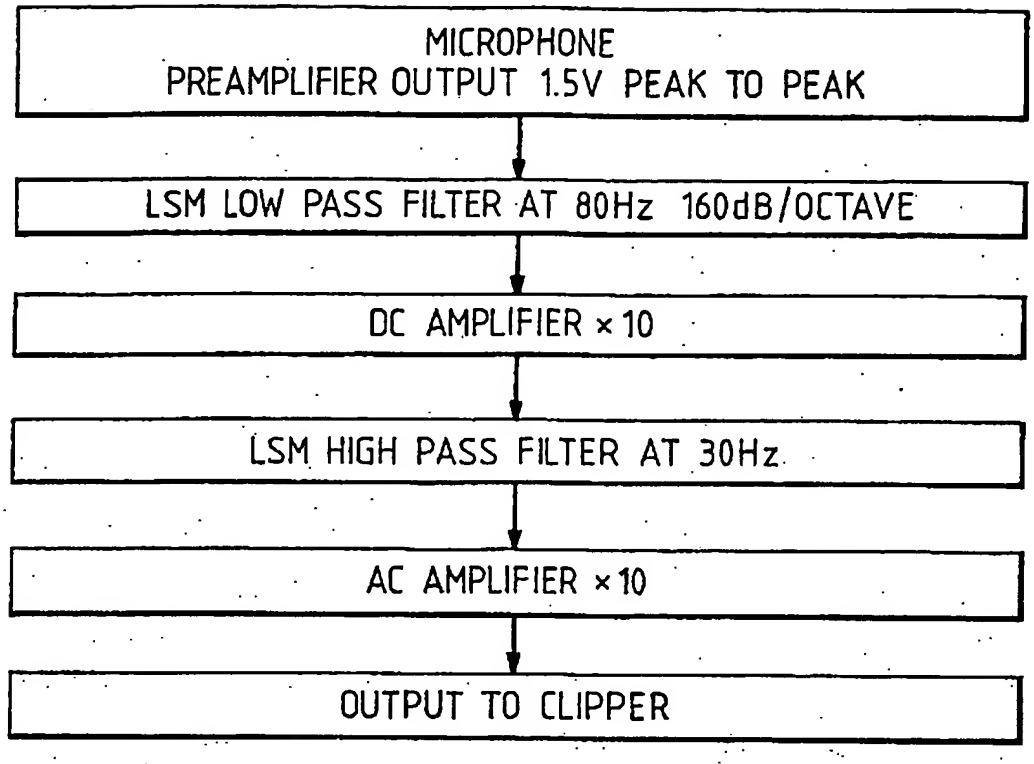
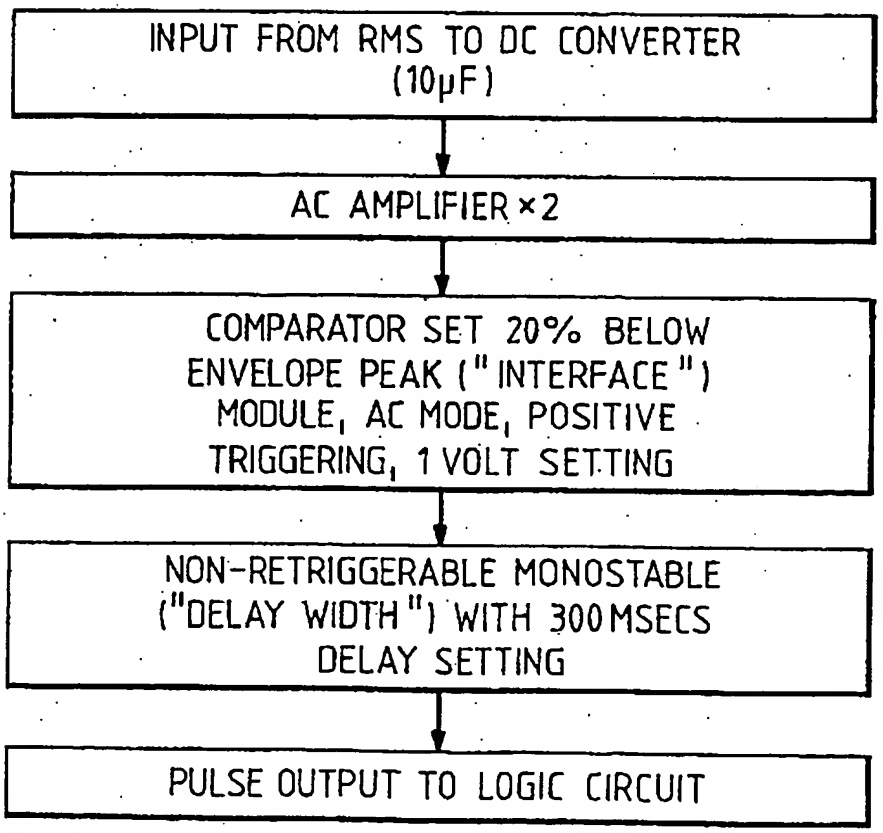

MICROPHONE
PREAMPLIFIER OUTPUT 1.5V PEAK TO PEAK
LSM LOW PASS FILTER AT 80Hz 160dB/OCTAVE
DC AMPLIFIER × 10
LSM HIGH PASS FILTER AT 30Hz
AC AMPLIFIER × 10
OUTPUT TO CLIPPER

Fig.8.


INPUT FROM RMS TO DC CONVERTER
(10μF)
AC AMPLIFIER × 2
COMPARATOR SET 20% BELOW
ENVELOPE PEAK ("INTERFACE")
MODULE, AC MODE, POSITIVE
TRIGGERING, 1 VOLT SETTING
NON-RETRIGGERABLE MONOSTABLE
("DELAY WIDTH") WITH 300 MSECS
DELAY SETTING
PULSE OUTPUT TO LOGIC CIRCUIT

SPECIFICATION

Cardio-respiratory monitor apparatus and method

The present invention relates to a cardio-respiratory monitor and apparatus for monitoring cardio-respiratory action.

The occurrence of apnoea in humans, particularly infants, and its detection has become a matter of some importance. Particularly in the case of infants, detection of apnoea may be used to prevent or reduce the occurrence of Sudden Infant Death Syndrome (commonly called "Cot Death").

Several monitors have been designed for detecting apnoea but they have hitherto relied upon detection of respiratory movements or movements of the body. These have frequently been in the form of pressure detectors which have either been placed under the body to detect movement of the body or attached to the chest to detect movement of the chest, the variation of pressure within a pressure capsule being used to provide an indication signal of respiratory function.

The present invention provides a method and apparatus for detecting not only respiratory function but also detecting cardiac function.

The present invention provides, according to one aspect, a cardio-respiratory monitor apparatus comprising a sound detector for detecting sound from a body, means for passing a signal representing the sound detected to an analyser apparatus, said analyser apparatus including means to distinguish the signal relating to the respiratory function and the signal relating to the cardiac function, and means to monitor these separate signals. It has been determined by experiment that, in general terms, both cardiac and respiratory function produce sounds which are of different frequencies which are separate from one another. In practice the cardiac function produces sounds of a frequency of approximately 30 to 180 Hz and the respiratory function provides sounds of a general frequency range of 200 to 950 Hz.

The monitor apparatus therefore includes means to separate these frequencies and to monitor them separately.

The sound detector may comprise a microphone including means to attach the microphone to the chest wall. The means to attach the microphone to the chest wall may include means for providing a space between the microphone and the chest wall and in a preferred arrangement this may be in the form of adhesive "O" rings.

In the analyser apparatus the monitor means for monitoring the two sound related signals may include a phase locking means in which, for example, in the case of the cardiac monitor, will analyse the rate of heart beat and will note any missing heart beats.

The present invention provides a method of monitoring life function comprising detecting sound from a body, analysing a signal derived from the sound detected to provide two output signals, one of which relates to the sound signal relating to respiratory function and the other of which relates to the sound signal relating to the cardiac function, and monitoring these separate signals.

The present invention also provides a method of determining life function comprising detecting sound from the chest of a human body, determining cardiac function from a first frequency range of detected sound and determining respiratory function from a second frequency range of detected sound.

Although through the specification we refer to the use of the monitor apparatus with human beings it will be understood of course that the apparatus may be adapted for use with other mammals and may have particular use in veterinary practice.

Preferred arrangements of the invention will now be described by way of example only with reference to the accompanying drawings in which:

Figure 1 is a diagrammatic perspective view of a cardio-respiratory monitor apparatus according to the invention,

Figure 2 is a block diagram showing the process steps in a first stage of the analysis of the respiratory signal,

Figure 3 is a block diagram showing the process steps in a second stage of the analysis of the cardiac signal,

Figures 4A to 4E show signal waveforms in the process steps of the analysis of the respiratory signal,

Figure 5 is a block diagram of the process steps in a third stage of the analysis of the respiratory signal,

Figure 6 is a block diagram of the process steps in a fourth stage of the analysis of the respiratory signal,

Figure 7 is a block diagram showing the process steps in a first stage of the analysis of the cardiac signal,

Figure 8 is a block diagram showing the process steps in a second stage of the analysis of the cardiac signal.

Referring to Figure 1, the monitor apparatus may be used with respect to a human subject 11 which may be an unsedated infant sleeping in a sound proofed cot in a quiet room. The sounds transmitted through the chest wall 10 of the human subject 11 as a result of breathing and cardiac action respectively are detected by a single sub-miniature sensor 12 consisting of a combined condensor microphone and vibration detector applied to the chest wall 10. We prefer to use a sub-miniature electret condenser microphone (Knowles Electronic Co., UK) housed in polytetrafluorethylene shells and applied to the chest wall 10 by adhesive "O" rings. The sensitivity of the microphone is 10 mV/Pa and the frequency response is uniform between 40 Hz and 3 kHz. In some instances two microphones may

be used in which case the gains of each microphone/amplifier combination should be equalised before measurement. Before use an absolute sound level calibration of 94 dB re 20 uPa at 1 kHz is also performed.

The electrical signal from the sensor 12 is passed by cable 13 to a monitor unit 14 containing electronic circuits, control and display panels. Within the monitor unit 14 the electrical output from the sensor 12 is amplified by amplifier 16 and divided into each of two high quality precision electronic filters 17, 18. The first filter 17 is set to pass waveforms with frequencies common to the cardiac sounds, that is between 20 and 200 Hz with an attenuation rate at each limit of 24 dB/octave. The second filter 18 is set to pass waveforms with frequencies common to the respiratory sounds; thus filter 18 may be a highpass filter with a lower cut off frequency of 180 Hz and an attenuation rate of 48 dB/octave (Barr & Stroud Variable Filter EF3 UK). The output of each filter unit 17, 18 is phase locked and may be finely tuned externally by control 48.

If desired the filtered sounds may be recorded on a seven channel FM tape recorder at 38 cm/sec. During recording sound quality is monitored through an audio amplifier and headphones.

If however it is desired to analyse and monitor the cardiac and respiratory activity in real time the cardiac and respiratory signals on lines 21, 22 are respectively rectified in rectifiers 23, 24 and passed to respective means 26, 27 which pass the signals through preset amplitude and time windows which are used to generate pulses at the start of every first heart sound and towards the end of every inspiratory phase of respiration (ie towards the end of every inward breath).

The pulses from window means 26 which signal the occurrence of the first heart sound are: 1) counted electronically by counter 28 and displayed on a front panel light-emitting diode display 29, and 2) used to activate a hear rate trend counter 31 with an external output to a visual display unit 32 or plotter 33.

The pulses from window means 27 which signal the occurrence of respiration are 1) counted electronically and displayed on a front panel light-emitting diode display 37, and 2) used to activate a respiratory rate trend counter 38 with an external output to a visual display unit 39 or plotter 40.

The pulses from window means 26, and 27 which signal the occurrence of breathing and cardiac action also pass to their respective but interconnecting logic alarm circuitry 41. In general, an alarm 45 is activated by the occurrence of a predetermined delay in either the cardiac and/or respiratory signals. The exact combination of alarm modes can be specified on the front panel controls 42.

In the normal baby, frequency analysis has shown that the principal heart sound components lie within the band width of 30 to 180 Hz and the principal respiratory components within the range of 200 to 950 Hz.

The monitor is intended for medical use principally in babies (including those born prematurely), infants and children but is also suitable for use in adults. The general age range of the patient is specified by the operator from an external control switch 43 on the front panel. Operation of the switch 43 has the effect of altering the filter and other window characteristics to the specific pattern of the heart and breathing sounds of the particular age range selected. Fine tuning is again possible if necessary in an individual case.

The occurrence of noise is specifically eliminated in the system by:

1) the use of narrow and specific frequency, amplitude and time windows.
2) phase locking of the specific cardiac and respiratory signals.
3) provision in the logic alarm circuitry for the isolation of atypical pulse surges.
4) background noise may also be elimnated by signal averaging techniques during frequency analysis or by subtraction of the simultaneously recorded background noise frequency spectrum.

Where the normal respiratory sounds are complicated by the addition of high frequency polyphonic wheezes (as in asthma) or variable frequency crackles (as in some obstructive and restrictive lung conditions), these events are detected by subsidiary filters 46, 47 engaged by the operation of a front panel selector switch 48. The amplitude of the wheezes or crackles on a breath-to-breath basis and/or the trend of amplitude with time are available as an external output to a visual display unit 51 or plotter 52.

A specific microprocessor based extension unit 54 may be attached to an appropriate output socket 52 of the monitor unit 14. With this addition, it is possible to generate values of tidal volume, flow rate and airways resistance from the primary input signal filtered to yield the respiratory waveform and digitised. In this case, the operator keys in the patient's body proportions by control 56 to the microprocessor extension unit and a continuous output of the selected variable or variables is available for display on a visual display unit 57 or plotter 58.

Although we have shown a separate VDU 32, 39, and 51 for each separate function and a separate plotter 33 40, 52 for each separate function it will be understood that these may be combined as a single unit if desired. The single VDU then being switchable between modes in which it will show all of the signals simultaneously, or each of the signals separately and similarly, the plotter can be switched between modes in which it will plot all of the signals simultaneously (there being provided sufficient pens for the number of input signals) or a single signal at a time.

We shall now examine the various parts of the

arrangement described with respect to Figure 1 in more detail.

The respiratory signal

Both inspiratory and expiratory sounds are transients consisting of random noise over a bandwidth 200 to 900 Hz. The resonant frequency is above 400 Hz. Expiration is less intense a sound than inspiration and may not be detected at all in some breaths. The mean peak sound pressure level of the inspiratory sound is around 65 dB re 20 μPa. The inspiratory sound normally lasts 300 to 400 msecs and the expiratory sound is shorter. There is a variable break in the sound between the two phases of respiration. Sound intensity is related exponentially to air flow rate but these are related linearly over the usual operating range. Breaths whose sound level falls below the threshold of detection are usually associated with inadequate ventilation. However, slow deep breathing in quiet (non-REM) sleep may pose a problem to detection in some circumstances even though alveolar ventilation is satisfactory. Instantaneous (breath-to-breath) respiratory rates are highly variable (eg 20 to 120/min). The actual respiratory rate over one minute does not normally exceed 80.

The cardiac signal

A single heart beat generates two sounds related to sequential cardiac valve vibrations. These are discrete, none random transients and (unlike the breathing signal) are relatively uniform from beat to beat. The resonant frequency is below 100 Hz. The mean peak sound pressure level is 30 to 40 dB greater than that of the respiratory sounds. The first heart sound last around 70 msec and is usually longer and more intense than the second sound. The interval between heart sounds corresponds to the interval between the Q and T waves of an electro-cardiogram ie approximately 150 to 200 msecs. The upper limit of hear rate is 180/min.

The microphone

The microphone 12 is a Knowles CA series insert which uses an electret film and contains an integral FET amplifier. It is used as a combined contact condenser microphone and direct vibration detector. The microphone 12 is centrally in contact with an air space of <5 mm^3, sealed by the skin surface of the chest 10. Peripherally, it is in contact with a 1 cm diameter PTFE plate 1 mm thick. This plate forms the contact surface of the microphone housing which consists of a polytetrafluorethylene shell containing alternate layers of epoxy resin and either dense latex foam rubber or soft polymetric elastomer.

The microphone specification is:

Size: 7.2×7.2×4.9 mm
Bandwidth: 20 Hz—2 kHz
Sensitivity: 10 mV/Pa (Breathing signal 0.5 mV approx.)
Supply voltage: 0.9—20 V DC
Current drain: 0.05—0.1 mA
Nominal output impedance at 1 kHz: 1,700 ohms.

The output from the microphone 12 is preamplified to give a signal of 1—4 volts peak to peak. This may increase to 10V+ when the human subject 11 moves.

The monitor unit 14

The raw sound signal has to be handled electronically in a series of individual steps. Each step is contained within a single module. This allows independent adjustment of each individual step of the signal analysis to arrive at an optimal arrangement. Alternatively, however, the individual modules can be integrated into a single, compact system employing integrated electronic circuits.

In handling the signal for monitoring purposes, signal distortion and phase shift are not critical provided that sensitivity and periodicity are maintained. System delays of up to 250 msecs are acceptable.

Processing the respiratory signal

Initial amplification and filtering

The signal from the microphone 12 is processed in the manner shown in Figure 2. The signal from the microphone 12 is pre-amplified, passed to an LSM highpass filter having a cut-off frequency of 350 Hz, the signal is then passed to an AC amplifier to amplify it by 50 times, the amplifier signal is passed to an LSM lowpass filter having a cut-off frequency of 650 Hz, and the output from the lowpass filter is again amplified in an AC amplifier by 10 times.

In the part of the process thus far described the filter 18 therefore comprises both the LSM highpass filter, the LSM lowpass filter, and the AC amplifier.

Generation of Breath Sound Envelope

The steps in this part of the process are illustrated in Figure 3. The signal from Figure 2 is rectified in rectifier 24 and integrated with a predetermined sampling rate of 6 Hz. The integrator holds the peak value of each 166 msec sample. The energy present in each sample during a breath is adequate to provide a satisfactory sound envelope without a significant response to intervening attenuated heart sounds or background noise. The signal falls to the value of the noise floor (ie the noise level) between breaths.

To improve the signal-to-noise ratio further, ie to enhance the discrimination between the breath sound envelope and the unwanted noise integrals the output is passed from the integrator through a non-linear-gain amplifier. An "antilogarithmic" amplifier is used in which large signals are made larger and smaller ones smaller, but a squaring amplifier could be used (eg $3^2=9$ and $4^2=16$, so an initial signal-to-noise ratio of 4:3 (1.3) is improved to one of 16:9 (1.8)).

Setting the Primary Trigger Level

The sound "envelope" is now used for the remainder of the triggering process.

An amplitude limit is set so that a signal is detected but noise is not. A retriggerable monostable is used for this process such that when the amplitude of the sound envelope exceeds the preset threshold, a pulse is generated.

The sequence of event is shown in Figures 4A to 4E. Figure 4A shows a signal corresponding to the breath sound, Figure 4B shows the integrated envelope of that signal and Figure 4C the antilogged integral of the signal of Figure 4B. If the voltage limit is set to 10% above the noise floor then the signal of Figure 4D is produced and if this is then passed to the retriggerable monostable the pulse output of Figure 4E is produced.

From Figure 4E it will be seen that pulses coinciding with breaths of Figure 4A have been generated. The pulses start as the upward stroke of the sound envelope crosses the threshold value and end as the downward stroke crosses the same threshold level. The process is summarised in Figure 5.

Setting Subsequent Trigger Levels

The breaths of Figure 5 have now been converted to pulses of Figure 4E.

The upward stroke of each pulse represents the beginning of a breath and this point is now used as a trigger.

The pulse train nees to be "smoothed" still further to prevent inappropriate triggering. At a breathing rate of 80/min, breaths will, if completely regular, occur no more frequently than once every 750 msecs.So a further constraint can be imposed on the pulse train—that pulses should occur no more frequently than one every 750 msecs (0.75 Hz). However, babies do not breathe regularly and very fast periods may alternate with much slower ones. Even at an instantaneous (breath-to-breath) rate of 120/min, there would be one breath every 500 msecs, so in a practical sense the most reasonable constraint is that the pulses should occur no more frequently than once every 500—750 msecs. In setting the actual maximum pulse frequency, it is of no consequence if 5—10% of breaths are actually missed and "overridden" each minute as the purpose of the monitoring is apnoea detection rather than 100% accurate respiratory rate monitoring. Further, the vast majority of babies have average respiratory rates at or below 40/min.

The processes of this stage are illustrated in Figure 6 and the shape of the corresponding pulse for each stage is illustrated on the left hand side of Figure 6.

The output from this stage is passed to the logic alarm circuit 41 and also to counter 36.

We now deal with the signal processing for the cardiac pulse.

Cardiac Pulse Processing

Initial Amplification and Filtering

The stages of this initial amplification and filtering are illustrated in Figure 7. The output signal which has been preamplified at the microphone 12 is passed to an LSM lowpass filter having a cut-off frequency of 80 Hz, the output signal from the LSM lowpass filter is passed to a DC amplifier where it is amplified by 10 times. The output signal from this amplifier is passed to an LSM highpass filter having a cut off frequency of 30 Hz and the output signal therefrom is amplified by an AC amplifier by 10 times. The amplified output is passed to a clipper.

As before, the filter 17 of diagrammatic Figure 1 comprises the LSM lowpass filter, DC amplifier and LSM highpass filter and AC amplifier.

Optimising the Signal-to-Noise Ratio

The filtered output from the previous step is passed to an amplitude clipper module. This module effectively extracts the noise band between selectable amplitude limits and rejects it. The positive and negative parts of the signal are then "joined up" again and amplified to a constant level.

The sound envelope is generated by passing the signal from the previous stage to an integrated circuit in which the RMS (root mean square) value of the signal is calculated and expressed as a DC value. The time constant of the system is 120 to 250 msecs. This process converts the pair of spike signals from the previous step into an "M" shaped envelope in which the first peak corresponds to the first sound and the next peak to the second heart sound. This helps to smooth the heart signal and improves the signal stability. This step can be omitted but false triggering is more likely.

Generating the Trigger Pulses

This stage is illustrated in Figure 8. In a comparable manner to the breathing envelope, the cardiac sound envelope activates a non-retriggerable monostable when the upstroke crosses a preset amplitude limit, set 20% below the peak amplitude level of the envelope. Further smoothing is achieved by specifying the period for which the pulse generator cannot be retriggered using a second monostable module. The refractory period selected is 300 msecs, accommodating heart rates approaching an upper limit of 180/min.

From this step a pulse output is produced which is passed to the logic circuit.

Alternatively this step can be performed with two variable non-retriggerable monostables, the first of which delivers a pulse of 300 msecs duration triggered on the upstroke of the sound envelope and the second monostable delivers a TTL pulse on the upstroke of the first heart pulse.

Movement signal

Any movement of the body will produce an audio signal. This audio movement signal is

normally in the range of 1 to 4 volts RMS after amplification. Signal levels greater than 5 volts RMS indicate additional noise. Signals greater than 7 volts RMS indicate movement or vocalisation.

The raw sound signal is monitored by an RMS voltage window comparator with a time constant of 2 secs. When the input signal falls outside the 1—5 V RMS window, the output from the comparator goes high for signals >5 V RMS and low for signals <1 V RMS. The outputs from the comparator are available as DC levels and also operate appropriately coloured lights automatically.

The above process has therefore now produced a TTL pulse for each breath and heart beat and also a high DC level which indicates movement. The apparatus includes digital circuitry to analyse these rates of pulses and the trend and to cause an alarm to operate if the rates and trends are outside predetermined limits which may be adjustable for age, weight and other factors.

The apparatus described particularly with regard to Figure 1 is suitable for experimental or hospital use. However, a simplified version of the apparatus may be produced for home use in which expensive components such as the VDU plotter may be deleted, the apparatus simply producing an alarm if predetermined parameters regarding cardiac and respiratory rate are exceeded. In such a case, the apparatus can be produced more cheaply as the electronic components can be produced in integrated circuit form.

The invention is not restricted to the details of the foregoing example.

Claims

1. A cardio-respiratory monitor apparatus comprising a sound detector for detecting sound from a body, means for passing a signal representing the sound detected to an analyser apparatus, said analyser apparatus including means to distinguish the signal relating to the respiratory function and the signal relating to the cardiac function, and means to monitor these separate signals.

2. Apparatus as claimed in claim 1 in which said signal distinguishing means comprises means to distinguish the frequency of the sound detected by the detector and to separate the respiratory signal and cardiac signal on the basis of frequency.

3. Apparatus as claimed in claim 2 in which the respiratory signal relates to the part of the sound detected in the range of 200 to 950 Hz.

4. Apparatus as claimed in claim 2 or 3 in which the cardiac signal relates to the part of the sound detected in the range of 30 to 180 Hz.

5. Apparatus as claimed in any of claims 1 to 4 in which the sound detector comprises a microphone including means whereby in use the microphone is attached to the chest wall of the body.

6. Apparatus as claimed in claim 5 in which the means to attach the microphone to the chest wall includes means for providing a space between the microphone and the chest wall.

7. Apparatus as claimed in claim 6 in which the means for attaching the microphone to the chest wall comprises adhesive "O" rings.

8. Apparatus as claimed in any of claims 1 to 7 in which the means for monitoring the separate signals includes a phase locking means which will analyse the rate of pulses produced by respiratory and cardiac action and note any missing pulses.

9. A method of monitoring life function comprising detecting sound from a body, analysing a signal derived from the sound detected to provide two output signals, one of which relates to the sound signal relating to respiratory function and the other of which relates to the sound signal relating to the cardiac function, and monitoring these separate signals.

10. A method of determining life function comprising detecting sound from the chest of a human body, determining cardiac function from a first frequency range of detected sound and determining respiratory function from a second frequency range of detected sound.

11. A method as claimed in claim 10 in which the first frequency range is 30 to 180 Hz and the second frequency range is 200 to 950 Hz.

Printed for Her Majesty's Stationery Office by the Courier Press, Leamington Spa, 1984. Published by the Patent Office, 25 Southampton Buildings, London, WC2A 1AY, from which copies may be obtained.